Principle of operation and use of accelerometry in assessing the musculoskeletal system – a narrative review

Zasada działania i zastosowanie akcelerometrii w ocenie układu ruchu – przegląd narracyjny

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Key words
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Abstract
Accelerometry is a relatively new but promising method of gait examination. It is based on the usage of sensors which measure linear acceleration at a certain material point. The purpose of this article is to review the literature on the subject from the point of view of applying this technique in assessing human gait, its advantages and shortcomings and the reliability of measurement. Papers by various authors have been reviewed and their results compared. Research concerned detection of the phases and events of gait, calculation of gait parameters such as speed and step length, balance evaluation and the monitoring of physical activity. In order to verify the correctness of the collected data, it was compared with the readings of the VICON system, force platforms and special electronic walkways. An analysis of the literature resulted in the following conclusions: the advantages of accelerometry is the low cost of devices, their small size and mass and measurement which is not limited to the laboratory. The disadvantage is first of all the necessity to use cables, which makes it harder to conduct the long-term monitoring of physical activity. The method is reliable if the experiment is properly planned and carried out. The most important conditions are the proper location of sensors, tight binding to the body, the most accurate alignment of the anatomical axis with the measurement axis and the usage of a proper algorithm for data processing. The authors of the majority of papers consider accelerometry to be a reliable and useful method of analyzing the parameters of gait. At present, accelerometers are used mainly for examining the model of gait and assessing dysfunctions, as sensors in FES assisted walking in patients with dropped foot and during physical activity monitoring.

Słowa kluczowe
absorpcja wstrząsów, akcelerometr, chód

Streszczenie
Akcelerometria jest stosunkowo młodą, ale obiecującą metodą w dziedzinie badań nad chodem. Bazuje ona na zastosowaniu czujników mierzących przyspieszenie liniowe występujące w danym punkcie materialnym. Celem tego artykułu jest przegląd literatury pod kątem zastosowania tej techniki w ocenie lokomocji człowieka, zalet i wad oraz rzetelności pomiaru. Przeglądnięto prace różnych autorów i porównano ich wyniki. Badania dotyczyły wykrywania faz chodu, obliczania parametrów takich jak prędkość czy długość kroków, oceny równowagi oraz monitorowania aktywności fizycznej. W celu sprawdzenia poprawności zarejestrowanych danych, porównywano je z odczytami systemu VICON, platform dynamometrycznych oraz specjalnych elektronicznych ścieżek. Analiza literatury dostarczyła następujących wniosków. Zaletami akcelerometrii jest niski koszt urządzeń, ich niewielkie rozmiary oraz masa, a także brak ograniczenia pomiaru do wnętrza laboratorium. Wady to przede wszystkim konieczność stosowania kabli, co utrudnia długotrwały monitoring aktywności fizycznej. Metoda jest rzetelna, o ile eksperyment jest prawidłowo zaplanowany i przeprowadzony. Najważniejsze warunki to właściwe umiejscowienie czujników, zapewniające dobre przyleganie do ciała mocowanie, jak najdokładniejsze skoordynowanie osi anatomicznej z osią pomiaru oraz użycie właściwego algorytmu przetwarzania danych. Autorzy większości prac uznają akcelerometrię jako wiarygodną i przydatną metodę do oceny parametrów chodu. Obecnie akcelerometry znajdują zastosowanie głównie przy badaniu wzorca chodu i oceny dysfunkcji, jako czujniki FES u pacjentów z opadającą stopą oraz podczas oceny równowagi oraz monitorowania aktywności fizycznej.

The individual division on this paper was as follows: A – research work project; B – data collection; C – statistical analysis; D – data interpretation; E – manuscript compilation; F – publication search; G – grant and funding acquisition

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A SHORT ANALYSIS OF GAIT

Man's gait is a complex phenomenon. Its complete description requires the analysis of much data including: kinetic, kinematic, electromyographic. Accelerometric tests which are based on the testing of linear and angular acceleration[1] play within this evaluation and analysis of gait a significant role. The method is enjoying a constant increase in popularity thanks to advantages and benefits such as: the relatively low cost of equipment; the small size and mass of the sensors, which do not limit freedom of movement while walking while at the same time may be used anywhere – the testing is not limited to laboratory conditions; while direct measurement in 3D reduces the risk of error[2,3].

We describe human gait as a cyclical phenomenon which means the recurrence of the self same motor structures at time intervals. We may differentiate a phase of support and swing (transfer) for each of the limbs. The phase of support constitutes on average 62% of the gait cycle and covers the entire time of the limbs contact with the ground. This is comprised of five sub-phases: initial contact*, loading response, mid-stance, terminal stance* as well as pre-swing. The remaining 38% constitutes the phase of transfer, during which the limb has no contact with the ground. This is composed of three sub-phases: initial swing*, mid-swing* and *terminal swing.*

Initial contact is a short lasting event, in the course of which the heel comes into contact with the ground. There occurs a moment of force initiating the movement of sole bending as well as the eversion of the foot, which is impeded by the eccentric workings of the muscles: of the anterior, the extensor of the long toe, the extensor of the hallux as well as of the posterior. Next the centre of gravity starts to move in a direction extended to the front of the foot and the loading response begins. This is possible thanks to the bending movement in the knee joint, which is limited by, among other things, the quadriceps of the thigh to around 15°. The mid-stance *phase begins when the foot is lying flat on the ground.* At this time in the other limb there occurs a pushing away and transfer into the swing phase. The whole weight of the body therefore rests on the supporting foot. The moving forward of the centre of gravity forces the dorsal bending of the foot, eccentrically controlled by the very strong soleus and calf muscles, supported additionally by: the posterior, the long flexor of the toes, the long flexor of the big toe as well as the articular muscle – long and short. The progress of the terminal stance phase depends on the mobility of the straightened hip joint. The forefoot becomes the axis around which the extremity supporting the weight of the body turns. The raised heel results in a lessening of the plane of support. The soleus and calf muscles work eccentrically and, aided by the posterior and peroneal muscles, guarantee the stability of the lower ankle joint and the transverse joints of the tarsus of the supporting extremity. Preswing is the final subphase of the phase of support. The opposite extremity commences the period of support by both feet. This causes the sudden relieving of the foot and leads to a bending of the sole by 20° as well as a bending of the knee joint by 40°. At the moment corresponding to around 54% of the gait cycle there commences the energetic movement of the dorsal bending of the foot, which results in a pushing away from the ground and the commencement of the phase of transferring the extremity.

During the initial swing the hip bends by 20°, the knee by a further 20° (resulting in total to 60°), while the ankle joint dorsally bends in order to ensure the foot an absence of contact with the ground during transfer. Momentum is given to the extremity by the concentrically working flexors of the hip joint. The task of the midswing phase is the continued lifting of the extremity and foot above the ground. Movement forward is the result of the action of the force of inertia, hence this period is defined as passive. The bending in the hip joint increases to 25°, while the upper ankle joint through further dorsal bending returns to an intermediary position. The raising of the extremity ends the terminal swing *phase*, during which there occurs a complete extension of the knee with the maintaining of a 25° bend of the hip and an intermediary position of the ankle joint. The momentum of the extremity is stopped by the eccentric activity of the hip joint extensors and the flexors of the knee joint. In preparation for the approaching initial contact the muscles become active: the greatest gluteal and the great adductor muscle[4,5].

THE ARISING AND ABSORPTION OF SHOCKS

It follows to examine gait as a motor process within the context of the physical laws which describe it and to which it is subjected. Values such as velocity, acceleration or momentum are inseparably connected with gait. The simplest movement which may be set in motion by a material point is a uniform motion. We physically describe velocity in a rectilinear movement as an increase in distance over a unit of time: $v = \Delta s/\Delta t$ or as the first differential coefficient of movement in relation to time ($v = ds/dt$). It results from Newton's 1st Law of Motion that no force acts upon such a body or that the forces in action are in balance with each other.

Then when a resultant force acts on the body, differ than 0, this being in accordance with Newton's 2nd Law of Motion, such a body will move with a uniformly accelerated motion $a = F/m$. Acceleration may be equally described as an increase in velocity over a unit of time $a = \Delta v/\Delta t$, or presented as the second differential coefficient of movement in relation to time ($v = d^2s/dt$). For the body to move with such a motion the force acting upon it must be constant. In practice, however, for a body at rest to start to move with a uniformly accelerated motion there is needed a certain time period during which its acceleration will increase from 0 to the obtainment of the final magnitude. The acceleration will not therefore have a constant value, and consequently the resultant force will change. In describing this part of the movement (a so-called spurt or jolt), we must use the third differential coefficient of position over time $z = d^3s/dt$, or write it as a change in acceleration in time ($v = \Delta a/\Delta t$).

Indeed the presence of imbalanced forces (constant or variable), acting on the various elements of the human body, is the cause of shocks during walking. The most significant here is the force of the reaction of the ground during the impact of the heel in the initial contact *phase.* The absorption of these forces is a key element in the mechanism of gait [6]. During the contact of the heel with the ground at the end of the phase of transfer there occurs a short-lived force connected with the change of momentum. The shock brought about by the reaction of the ground is transferred through the skeleton to the whole body[6,7]. This impulse is short and lasts usually around 10–20 ms, while its magnitude is individual for each person. This value also depends on the speed of gait, the type of footwear, the ground surface, and even on the mood of the person in question. Tests confirm that shocks brought about by the reaction of the ground are dangerous for one's health. They are considered to be the main cause of the degenerative changes within the joints, headaches, the loosening of endoprostheses, inflammation of the fascia of the sole and the Achilles' tendon, muscle damages and compression fractures[6]. When during gait the foot hits the ground, in a similar way to a moving object hitting a stationary one, there occurs between them an exchange of energy and momentum. The momentum appears in two forms: linear, as a product of mass and velocity $p = mv$; as well as angular (the moment of momentum) being the product of the arms of force and momentum $\vec{L} = \vec{r} \times \vec{p}$. The principles of the conservation of momentum and the moment of momentum state that they cannot be created or destroyed. Momentum and the moment of momentum may be transferred from one object to another. And therefore the hitting of the ground by a heel in motion results in a transfer between them of momentum – the momentum of the Earth grows by that value that the momentum of the foot decreases. On the scale of the planet this is obviously a minute value. In a situation where the material located between the foot and the floor has an elastic property there occurs a return of momentum to the foot, which results in an increase in the total momentum exchange. This means in practice that the more elastic the ground the greater the momentum exchange and consequently the greater the value of vibrations produced in the foot.

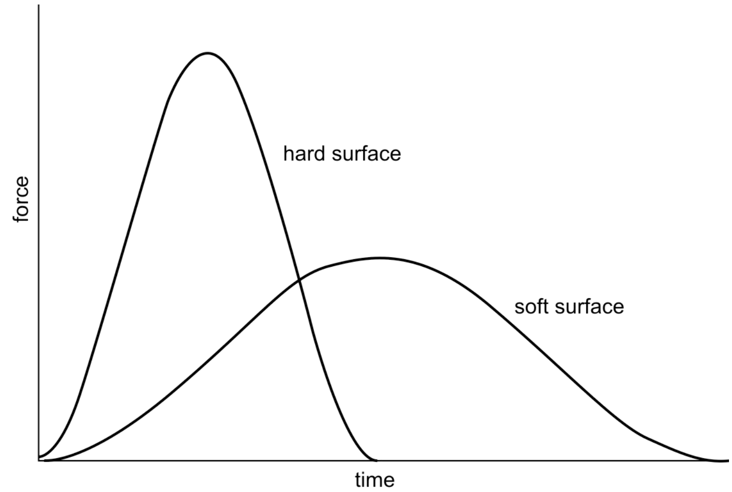

Figure 1
Amount of force and its duration during contact with soft and hard surface

The heel, finding itself in motion, possesses kinetic energy described by the formula $E_k = 1/2mv^2$. During impact with the floor, a part of this energy is transferred to the Earth. The remaining part of the energy is lost in the form of sound and heat. A bouncy floor which possesses the ability to preserve/hold it and to return this to the foot reduces these losses.

The magnitude of the shock wave which dissipates over the whole body as a result of the impact of the heel on the ground is proportional to the value of momentum exchanged between the foot and the ground. This is connected with the momentum change of the foot. This depends consequently on two factors: the magnitude of the momentum, which is the product of the mass and velocity of the foot (reduced prior to impact during the terminal swing *phase*) as well as the time during which this exchange in momentum takes place. The latter is influenced by: the anatomical construction of the foot, the sole and insole of the shoe as well as the character of the ground. The longer the route the heel bone has to cover during the entry into contact with the ground, the longer the time during which this takes place. In effect the forces as equally the impact wave brought about by them become respectively smaller (Figure 1).

Recapping, the forces arising in the contact of the heel with the ground are dependant on two properties connected with the foot (the mass and velocity) as well as on three connected with the ground (the thickness, elasticity and plasticity):

- the mass generating the shock wave – according to tests is on average ≈3.6 kg[8] – is greater than the mass of the foot itself but less than the whole of the lower limb. In practice the percentage of the mass of the entire limb that participates in the creation of the shock wave is ontogenetically variable;
- the velocity of the heel at the moment of contact with the ground; its value equally is individual – a part of individuals brake the foot almost completely in the air, others hit the ground with it at full speed;
- the thickness of the sole between the heel and the ground, which allows for an extension in the time of impact with the ground; a shorter distance to be covered shortens the time necessary for a reduction in velocity and in effect significantly increases the force of impact;
- the elasticity of the ground may increase the magnitude of the impacts/shocks; a highly elastic ground surface takes in energy lost through the impact of the heel and then returns it to the limb;
- ground plasticity is its ability to deform; an example may be a steel ground surface, which bends/yields insignificantly as a result of heel impact, while mud or sand subsides, extending the time for the holding of the motion; the material used in shoe insoles should combine features of plasticity and elasticity; pressed down under the weight of the body the insole, after the rising of the leg, returns to its

initial thickness and is again ready to fulfil its function during the subsequent step [6].

When the foot moving with a given velocity strikes the ground there is created the force of the ground reaction which results in the stopping of the heel. In accordance with Newton's 2nd Law of Motion ($F = ma$) the appearance of a force acting on a given body is connected with the acceleration of that body. Acceleration is brought from the heel, through the upper ankle joint into the tibia. Subsequently through the knee joint to the thigh bone and the hip and upwards through the spine right up to the skull. This phenomenon is known in the relevant literature as the shock wave. If an object moves downwards the force of the reaction of the floor therein created results in a slowing down of this movement [6]. According to research the tibial bone is, in the initial contact phase, exposed to an excessive load of a value of around 80 ms^{-2} (8g), at a time when this is much less on the skull[9].

There exist at least two mechanisms protecting the organism from the damaging action of forces bringing about shock waves. Their task is the absorption and dissipating of shocks (impacts). The first is comprised of the relevant movement parameters within the joint as well as its arrangement (for example bending of the knee[7]). The second is the correct anatomical construction – the presence of materials in the sole[10] and joints of viscoelastic characteristics as well as bone build. There exists a positive correlation between the mass of the bones and the weight of the body. A greater body weight represents greater forces of reaction in contact with the ground. An increased bone mass represents a greater bone section field and with it a greater surface area on which the forces unfavourable for the organism are spread [6,11]. An important element is also the significant smoothness without sudden impacts of the heel upon the ground.

The ground's vertically placed forces of reaction occurring during a person's locomotive activities are directed vertically upwards and dissipate outwards from the lower limbs through the torso to the head. For this reason the mechanisms alleviating the impacts must be located along their path. An important role is here fulfilled by the joints and muscles. The joints possess elastic-plastic properties. This is ensured by their appropriate construction and the physical properties of the materials within their composition, i.e. hyaline cartilage, synovial fluid and ligaments.

The muscle system is involved in shock absorption in two ways. The first is the superiority of the muscle tone of the flexors in relation to the antagonists. The second is the muscle cuff around the joint. It is worth drawing attention to the two main areas for the localisation of shock absorption. The first area is the foot supported on a semicircular base. This begins from the calcanean tuber, stretching through the side edge of the foot and the metatarsal-digital joints and ending at the metatarsal-digital joint of the hallux. The foot itself also constitutes a bone-joint chain stretching from the calcanean tuber to the toe joints. They therefore create the bridge of an elongated and transverse arching of the foot. Thanks to its springy-elastic properties it constitutes the first and the most important defence mechanism against shocks. The second area constitutes the system of spinal curvature. This is constructed of 24 bone segments arranged in series constituting an unquestionable defence against shocks/ impacts during a person's locomotive activities.

THE CLINICAL APPLICABILITY OF ACCELEROMETERS

Accelerometers started to be used in tests in the 1970s. In 1973 Morris[12] as one of the first proposed the evaluation of body movements by means of accelerometers. However, their improvement and refinement has occurred only during the last 10 to 15 years[13]. Thanks to present technology it is possible to produce small sized accurate instruments[13].

The use of accelerometers in the analysis of gait brings with it many advantages: the low cost of the devices, the lack of a need to restrict tests to laboratories, the small size of the devices, unlimited freedom of movement, the possibility of constant monitoring of movement activity during the normal day as well as the ability for direct 3D measurements reducing the number of mistake connected with differentiations in position and velocity[2,13,14,15,16,17]. Accelerometers are not, however, devoid of drawbacks, this is rather something dealt with in the relevant literature[2].

The basic question influencing the correctness of measurement is the attachment of sensors measuring acceleration to the patient's body. The position should be chosen after having taken into consideration the subject of the test. For example in testing gait the sensor is often placed on the shin or ankle; positioning on the wrist is used in measuring shakes in Parkinson's Disease. The best place in the monitoring of the activeness of the whole skeletal-joint system is a position close to the centre of gravity. The locating of the sensor in a concrete place of the tested area also significantly affects the exactness of the testing. For example, positioning it too close to the axis of rotation results in an ineffectual gauging of the amplitude values of the movement tested[13].

Accelerometers are attached to the patient's body most frequently by means of elastic bandages and straps ensuring a stable adhesion to the skin. Of interest is that sensors thus placed display a somewhat overestimated amplitude in relation to the sensors attached to bones[2]. The attaching of accelerometers to the skeleton is, however, an invasive method and for obvious reasons is rarely used in tests.

The most commonly practiced application of accelerometers during gait is their use with patients with dropped foot as an FES sensor[18,19]. Classic sensors do not give totally satisfactory effects chiefly as a result of the ineffective detection of heel contact with the ground (it occurs that they detect it during the transfer phase when small forces appear, ones acting on the sole; these are also ineffective with patients dragging a foot across the ground) as well as the possibilities of implanting a patient with them (patients often reject systems which require a lot of time in attaching or which look unsightly)[18]. A solution to the problem could be the use of miniature accelerometer sensors.

Usually accelerometers are used to evaluate general parameters of gait[20], in the diagnosis of symptoms such as various movement disturbances, disturbances in gait, shakes and in the hand trembling[21,22], the evaluation of the risk of fall in elderly people[23] or the monitoring of physical activeness[21,24,25,26,27,28]. The literature cites other interesting examples of the application of accelerometers in clinical practice like, for example, using it to uncover cases of epilepsy[29] or in the evaluation of swallowing in patients with dysphagia[30].

The key to aiding gait (for example in FES) is the correct differentiation of the particular phases of its cycle in time. In the literature there are many works whose authors, in applying various algorithms and methods, have attempted as exactly as possible to determine the main course of gait, amongst which the most important is the moment of the heel touching the ground (IC – initial contact) in the phase of transfer to that of support, as well as the detaching of the toes (EC – end of contact) at the transfer of the phase of support to that of transfer.

THE AIM OF THE WORK

The aim of the present work is:

1. an overview of the subject literature and the presentation of the results of various authors utilising an accelerometer as a method for research on gait;
2. a defining of research exactness and reliability as the factors on which these depend;
3. a giving of the most frequent practical uses of the described method.

MATERIALS AND METHODS

The databases of Science Direct, PubMed/Medline as well as the Physiotherapy Evidence Database (PEDro), were reviewed with a manual search subsequently conducted. The keywords *accelerometry*, *gait*, *gait event detection*, *acceleration gait analysis were used*. As a result of this 7162 items were found – of which 315 articles were selected, whose titles corresponded to the subject matter of the present article. After an analysis of the summaries a final number of 22 were used as a basis – of which 20 were research works and 2 systematic overviews.

The combined number of patients who took part in the tests covered by the various works was 300 - both healthy individuals as those with various illnesses. As a result of the varied character of the experiments (from tests on children through analyses of healthy adults to experiments with sufferers of Parkinson's Disease) the age range of those tested was very wide and fluctuated from 3 to 79.

Table 1

Accuracy in predicting IC events in the studies of different authors

Author	Number tested	Location of sensor	Exactness in measurement
Brandes[14]	20	Lower dorsum	99.6%
Zijlstra[31]	15	Lower dorsum	from 2±27ms to -103±25ms
Jasiewicz[15]	41	Dorsal side of the right and left shoe as well as the abdominal left and right surface of the lower leg/	from -11±23ms to 61±10ms
Gonzales[32]	11	L3	13±35ms
Hanlon[33]	12	Right ankle, right knee	-17±38ms
Lau[34]	13	Hip, tubercle of the tibia, shoe heel	PI: 0.954 – 0.987
Mansfield[18]	4	Lumbar area	98.2%-99.8%
Nienhuis[35]	26	Pelvis	-14±24ms

RESULTS

The determining of the moment of change of the phase of transfer into the phase of support and in reverse

The authors of the works cited in the current article[14,31,15,32,33,34,18,35] have concentrated in their research on identifying the moment of the heel's entry into contact with the floor. This element is essential both in functional electro stimulation as well as during the evaluation of gait itself. It enables one to differentiate on the graph obtained the particular cycles and allows one to calculate the length and number of steps, the pace of gait, its speed and regularity.

All the authors obtained very good results. Brandes[14] calculated the number of steps on the basis of determining the IC quantity. It was possible for a system based on data from the accelerator placed on the lower part of the back to calculate the number of steps with an accuracy of up to 99.6% in comparison with the factual state (recreated from video tape). Mansfield[18] in turn compared the accuracy of two devices – an accelerometer and a standard footswitch used for FES. The precision of the accelerometer is located within the borders of 98.2%–99.8%, while the precision of the footswitch was not significantly lower and was 92.4%–98.7%. Zijlstra[31] compared two different methods of determining the gait cycle on the basis of data from an accelerometer. The differences in time between them and the results read from the treadmill were from 2 ±27 ms to –103 ±25 ms, depending on the method and the speed of the treadmill. Jasiewicz[15] tested, by means of three different methods, the ability to predict IC in real time in a group of healthy individuals as well as in patients following injury to the spinal cord. In the healthy individuals all three displayed a similar measurement accuracy between: –11 ±23 ms, and 23 ±28 ms. In patients following injury the most accurate was the prediction of IC on the basis of the linear acceleration of the foot as well as the angular acceleration of the foot in the sagital plane (from –17±18ms to 28±31ms). Somewhat poorer results were given by the method of calculation based on the angular acceleration of the shin in the fibular plane where the error was from –15 ±17 ms to 61 ±10 ms. Gonzales[32], Hanlon[33] and Nienhuis[35] have obtained in their research respectively 13 ±35 ms, –17 ±38 ms and 14 ±24 ms of error in the determining of IC. Lau[34] has calculated for his research a PI indicator (in delimiting the general reliability and accuracy of measurement, its ideal value was 1), which for the identification of the beginnings of the phase of support and transfer is, respectively,: 0.987 (for the sensor on the shin) and 0.954 (for the sensor on the heel). All the above results are presented in Table 1.

The tracing of angular changes in joints by means of an accelerometer

Wong[36] and Tekada[37,16] attempted to calculate the changes in joint angles on the basis of data from an accelerometer. The correctness of the calculations was overseen by the VICON system. Wong's experiment[36] involved the measurement of the angle of trunk bending in the sagital and frontal plane. Measurement error constituted up to 5º for static conditions and up to 7º for dynamic conditions. Tekada[16] defined the angular positioning of the hip joint during bending/straightening and adduction/ abduction as well as of the knee joint (bending/straightening). The measurement error was respectively: 8.72 ±6.57º, 4.96 ±3.30º as well 6.79 ±4.65º. The same author conducted other tests[37], where he drew two graphs of angular changes in the hip and knee joint during gait: one based on changes in acceleration while the other on the VICON system. In the case of the hip the angles differed by 40–80%, although the peak values (appearing during changes in the direction of joint rotation) were similar. In turn the graphs of both systems for the knee were very close to each other but the maximum bending differed by 10º.

The weakness in the above cited works is the small number of patients tested – with only three volunteers taking part in each of them.

The use of an accelerometer in the evaluation of balance changes

Mayagoitia[38] has used an accelerator to evaluate the ability of a patient to maintain balance. The data was collected from an accelerometer placed on the back at the height of the centre of balance as well as from a dynamometer platform. The patients took up four different positions: standing slightly astride, the same with eyes closed, standing with feet placed together and with eyes closed. On the basis of the parameters of inclines of the centre of balance, based on algorithm data, both systems attempted to predict the patient's position. The accelerometer correctly differentiated the changes of position in 19 out of 20 tested; the platform was effective in 16 cases.

Motor activeness evaluation by means of an accelerometer

The small size of accelerometer sensors, the lack of limitation in their application beyond laboratory conditions as well as the long time for the registration of data open up possibilities for an evaluation of patient physical activeness during the course of normal day-to-day activities. Forester[21] conducted an experiment in which he checked the opportunity to differentiate a person's individual activities on the basis of the data of registered acceleration on the sternum, wrist and lower limb. Attempts to differentiate the following nine activities were carried out: sitting, standing, lying, sitting and talking, sitting and using a computer, walking, going up stairs, going down stairs and riding a bike. The research showed as many as 33% erroneous results. After the reduction to 5 in the number of the various activities (sitting, walking, lying, standing, riding a bike) the number of errors fell to 4.5%.

A differentiation in going up and down stairs was successfully achieved, however, by Italian researchers[39]. In the experiment conducted by them 24 volunteers took part (12 seniors and 12 young people). Those tested had to cover a track that led both inside and outside the building, composed of various surface levels, inclines set at various angles as well as steps (going up and down) of various heights. The data collected by a bi-axial accelerometer, was compared with the protocol in which were noted down in turn all the activities of those tested. The distinguishability of the above motor activities was high in both the younger and older testees and was, respectively: >90% and >92%.

Evaluation of other parameters of gait on the basis of acceleration data

In the aiding of a step by means of FES there is needed a precise determining of the moment the heel strikes the floor. It is, however, not always necessary during the usual evaluation of patient gait. Significant is, however, the differentiation of parameters such as velocity, pace, step length as well as its changeability. Hartmann[40] has attempted to determine the reliability of measurement carried out by a tri-axial accelerometer, comparing it with the results of the GAITRite system recognised by many authors as being reliable[41,42,43]. Excellent results were achieved – the coefficient of the ICC correlation for velocity, pace, duration and the length of a single pace was from 0.99 to 1.00. Somewhat worse results, although for that still very good ones, were obtained by Lord[44], in comparing the results of another system utilising accelerometers also with the GAITRite apparatus. The conformity of the parameters of gait was tested in individuals with Parkinson's Disease as well as in a healthy control group. The ICC for both groups was, respectively,: 0.92–0.99 as well as 0.76–0.95.

In the next experiment, Mayagotia[45] compared the kinematic parameters of gait obtained by an accelerometer, a gyroscope as well as the VICON system at five different gait velocities. The exactness of the data from the accelerometer clearly fell with the increase in velocity. At 1.4 km/h the measurement error was: 0.58 ±0.11 m/s^2 for linear acceleration of the knee and 1.58 ±0.27 rad/s^2 for the angular acceleration of the shin. This increased with the growth in velocity and for 4.6 km/h already constituted 2.16 ±0.59 m/s^2 and 5.37 ±0.66 rad/s^2 respectively. This was more than likely the effect of the impact and vibrations transferred to the metal strip of the accelerometer sensors, which occur during the impact of the heel on the floor. This fact did not have a greater influence on the gyroscope readings. The experiment also showed the weak sides of the VICON system. A large number of the measurements were not useable owing to the screening of one of the markers or other and the lack of the possibility to register its position by camera.

Result repeatability in accelerometer tests

Moe-Nilssen[46] has described an experiment he conducted, testing the repeatability of the results of an accelerometer. A portable, tri-axial accelerometer, placed on the lumbus spine, with the aim of balance evaluation while standing and gait. Nineteen healthy students were tested, who stood on both feet with their eyes open as well as on one foot with eyes open. The gait was tested by means of five randomly selected velocity on an even and uneven ground surface. The test was repeated after two days. The standard deviation was designated from the results of each testee and the internal class coefficient (ICC). The standard deviations showed a high repeatability, while the ICC for the majority of tests was contained in the values 0.79–0.94.

Similar tests were conducted by Henriksen[17]. The author describes the testing of 20 healthy individuals during which acceleration was measured with an accelerometer placed on the lumbus spine for six randomly selected gait velocities. The test was repeated the next day. After the converting of the initial data to the coordinating system the coefficient correlation as well as the level of measurement error was calculated. ICC had high values and this oscillated within the borders of 0.77–0.96; the measurement error was 0.007–0.01 g for acceleration, 0.009 m in step length, 0.022 m for the entire gait cycle as well as 1.644 step/min for the gait rhythm.

High indicators of measurement conformity have also been obtained by researchers from Australia in a test conducted on a group of eight people[47]. Accelerometers were attached to the patients' head, the neck of the lower trunk and the right shank. The patients took part in two identical sessions, in the course of which they had to cover five times a 30-metre distance at various speeds. The data from the accelerometers was computer processed and the coefficient of determination was calculated, the values of which fitted within the 0.60–0.99 band.

Also the authors of both the systematic surveys, Kavanagh[2] and Godfrey[13] emphasise the accelerometer as a reliable method for gait analysis, of changes in dynamic posture control connected with age as well as models of gait in individuals with motor disturbances[2]. They emphasise the already mentioned advantages of the devices, drawing attention, however, to such drawbacks as the limited practicality in conducting long-term measurements of patient activity resulting from the necessity of using many cables twisted around the joints and body[13].

DISCUSSION

Despite its relatively short history the accelerometer appears to be an interesting and promising method in the field of research into gait. Though it was known before its real development has fallen, however, within the period of the last 10-15 years. That said not much research has been conducted using this method. The authors base themselves the most frequently on analysis conducted by the VICON system, on the basis of which one can calculate the parameters of gait with great accuracy as well as determine its various indexes. Dynamometric platforms also enjoy recognition, treadmills as well as electronic gait tracks like, for example, the GAITRite device. These devices, although tested and precise, are for all that expensive, take up a lot of room and require a specific location; they cannot therefore be used outside of the laboratory, which severely limits their usage range. There also exists the risk of markers being shielded during movement[45], making the date collected during measurements useless.

In the above context the small, cheap and portable accelerometer is a very interesting alternative. However, for the device to fulfil its function as follows, it has to fulfil a series of fundamental conditions. Otherwise the result might not reflect reality, the consequence of which could be an incorrect diagnosis of the problem under test.

The first aspect is consideration of the subject of the research and the appropriate localisation of the sensors. In the case of long-term measurements of physical activeness or the use of the accelerometer as a FES sensor one cannot forget about patient comfort. The apparatus must be installed in such a way that it does not restrict movement and so that it does not have an unsightly appearance. Another element is the appropriate fastening of the sensor. The simplest method, one allowing for the best results, turns out to be the use of elastic bands enabling the accelerometer to be tightly attached to the body. One may presume that the placing of sensors on large muscle groups (like, for example, the calf triceps muscle) results in an absorption of vibrations, a reduction in amplitude, and as a consequence to a non-evaluation of results. In this case it would be better to place the meter close to the shin bone. In practice, in studying the relevant literature, one may note that the majority of authors draw attention to the fastening of the accelerometer in the location of hard skeletal elements such as: the vertebral body, the sternum, the lateral ankle, the iliac ala or the tubercle of the tibia.

It follows to remember that gravitation is a constant component in accelerometer measurements within the vertical axis. The result will therefore comprise a static part (gravitation) and a dynamic part (body movement). As a result of the impossibility to match ideally the anatomical axes to the axis of measurement it will be impossible to exactly separate these two components. However, the deviation between the axis of measurement and the anatomical axis is considered constant depending on the velocity. The obtainment of accurate values is therefore possible only thanks to the mathematical correction of the data[17].

The weakest results were obtained by authors attempting to measure variable angular values in the joints. In one of these the graph of these changes was shifted in time in relation to the graph of the VICON system by 40–80%, while the peak values of the angles of the knee joint differed by 10°[37]. After an improvement of the algorithm and a change of sensor position, the same researcher was able to obtain much better correlated results[16], while the measurement error in relation to the movement range[48] reduced from being 7.1 to 3.1 percent. In both experiments gyroscope sensors were used in addition to accelerometers. A measurement error of the order 5–7° for movements of bending and straightening the spine in the sagital and frontal plane[36], may be considered a good result.

While conducting measurements it follows to remember that the muscle-skeletal system creates an artificial mechanism for the absorption of shocks[11,6]. Therefore one should also expect greater amplitudes in the lower parts of the body than in those situated further up[49,50]. Of significance during the conduction of measurements of rotatory movements is the distance of the fastened linear accelerometers from the joint rotation. The further the sensor is from the axis the greater one should expect the measured linear velocity to be, and with this the acceleration must be greater.

The testing of the parameters of gait, such as velocity, the number of steps, their frequency as well as length gave good or even excellent results. Here the effectiveness of accelerometers was never questioned. The results of the works of Hartmann[40] or Lord[44], whose ICC fitted respectively within the borders 0.99–1.00 and 0.76–0.99, points to the high degree of method reliability. Equally in the works by Henriksen[17] the measurement errors were minimal.

Also the authors, in attempting to define the moment of initial contact of the heel with the ground, obtained good results. In a large part of the works sensors were exclusively placed on the torso, close to the centre of body mass, which meant that on the basis of its movements one could determine the actual phase of gait with a high degree of accuracy[14,31,32,18,35]. The differences were in general several hundredths of a second. This is a particularly significant criteria in the potential application of an accelerometer as a FES sensor.

Accelerometer sensors have turned out to be reliable in the evaluation of balance. In the research of Mayagoitia[38] their effectiveness in detecting patient positions was 95%. However, the detecting of patient motor activeness requires thought and their appropriate division according to dynamics. Problematic turned out to be the differentiation between such activities as: sitting, sitting and talking, sitting and using a computer; as well as going up and down stairs. After the modification of this division the effectiveness of model identification increased from 67% to 95.8%[21]. In turn other researchers were able to correctly distinguish walking on a level, on sloping surfaces as well as going up and down stairs with an accuracy of 90 to 92%[39].

CONCLUSIONS

The presented review of the subject literature allows one to draw the following final conclusions:

1. The accelerometer is a good method in the evaluation of gait as a result of the non-invasive nature of the devices, the lack of limitation to laboratory conditions, its sensitivity and exactness in measurement.
2. Its reliability depends on the proper determination of research aims, the use of an algorithm for processing the data, the adapting of sensor location to the type of parameters tested, the care with which they are fastened as well as the correlation of the registered axis with the anatomic axis.

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